A QUANTITATIVE STUDY OF THE MECHANICAL BEHAVIOR OF ENDOPELVIC FASCIA

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By

Richard Trapnell Hart

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A QUANTITATIVE STUDY OF THE

MECHANICAL BEHAVIOR OF ENDOPELVIC FASCIA

Approved:

Hyland Y. L. Chen, Chairman

Raymond P. Vito

Stephen L. Passman

Date approved by Chairman 5/23/77

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DEDICATION

To my Mother, Father, Sister, and Brother

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Summary

Richard Trapnell Hart

A Quantitative Study of The
Mechanical Behavior Of Endopelvic Fascia

54 pages

Directed by Dr. H.Y.L. Chen

A one dimensional materials testing device was designed and constructed in order to test two tissue samples of endopelvic fascia - one from the pubourethral ligament and the other from the dense regular connective tissue from the pelvic sidewall. Both tissues were tested in simple elongation, stress-relaxation and ultimate strength. The specimens were found to fit reasonably well the viscoelastic models of Fung. The ultimate strength test suggested the possibility of a localized failure in the endopelvic fascia resulting in large deformation of the pelvic supports and resulting in the symptoms of pelvic relaxation.

CHAPTER I

Introduction

Pelvic relaxation is a general term used to describe all conditions in which the supporting structure in the female pelvis no longer maintains the pelvic organs in their normal position. This would include uterine prolapse, cystocele, rectocele, entrocele, and the various combinations of these. It is estimated that 100,000 to 125,000 operative procedures are performed for pelvic relaxation each year.

The economic impact of this can be projected. Recent figures from the Crawford Long Hospital estimate that the surgical treatment of pelvic relaxation costs approximately $2500. (A breakdown on this would be: surgeon's fee- approximately $450, hospital bill- approximately $1250, and six weeks loss of employment figured at $135 per week would be about $800.) This represents an annual expenditure of $250,000,000 in this country alone for the surgical treatment of pelvic relaxation. In spite of this, a careful search of medical writings reveals a paucity of research into the causes of

pelvic relaxation.

The prevalent hypothesis is that the endopelvic fascia undergoes a generalized relaxation which results in a stretching or attenuation of the endopelvic supports. When the stretching becomes extensive and the fascia herniates, surgical repair is normally indicated. As a consequence of the prevalent theory, operative procedures have been devised to plicate and shorten the elongated fascia. But, despite the general improvement in operative techniques, the five year cure rate is disappointingly low (60-75%). In fact, there is a poor understanding of the anatomy of the endopelvic fascia and virtually no detailed description of what happens to the fascia when pelvic relaxation occurs. Recently, twenty standard textbooks of gynecology were reviewed and on this subject, only two presumptions were offered: (1) pelvic relaxation resulted from some form of "childbirth injury", and (2) consequently, there occurs a general attenuation or a generalized stretching of the endopelvic fascia. The only evidence in support of "childbirth injury" is the predilection for the occurrence of pelvic relaxation in women who have borne children by vaginal delivery. Paradoxically, it is seen occasionally in women who have never been pregnant. A further paradox is that it is rarely closely associated in time with

childbirth, but usually occurs in the perimenopausal years, approximately ten to fifteen years after the birth of the last child. As to the general attenuation of the fascia, no evidence has been accumulated to document this idea. It might be mentioned that rarely anywhere else in the human body does a generalized attenuation of connective tissue occur. Usually connective tissue only tears or stretches locally -- it does not uniformly attenuate.

During the last five years, 93 operations have been performed in the clinics of Dr. A. C. Richardson in which demonstrable tears in the endopelvic fascia were repaired. These were not general plication procedures and had an overall cure rate of over 95% [8]. In 3 of the 4 cases where Dr. Richardson's technique failed, the failure was evident within 6 weeks after surgery. This failure rate contrasts with the failure rate of the prevalent operative proceedures which continues to increase over time. Thus, it seems that the phenomenon of pelvic relaxation is the localized failure or tearing of the endopelvic fascia and not a general attenuation of the fascia.

In order to help clarify the true cause of pelvic relaxation, the mechanical properties of the endopelvic fascia must be determined. In order to do so, a

one-dimensional materials testing device, named "Alphatron" after Fung's constant alpha [3], was designed and constructed in the Biomechanics Laboratory of the School of Engineering Science and Mechanics. The Alphatron enables quantification of the mechanical behavior of the connective tissue composing the structural support in the pelvis. Two tissue samples of endopelvic fascia were made available by Dr. A. C. Richardson for testing. The results from these tests lay the foundation for further studies [6].

Appendix I contains information about the design, construction, and use of the Alphatron. Appendix II contains a partial bibliography for pelvic relaxation which may be useful for further studies of pelvic relaxation.

CHAPTER II

Method

Experimental Apparatus

The purpose of the experimental system is to provide a means for one-dimensional tests of soft tissue while monitoring the time course of force and stretch to deduce the stress-strain history relationship.

The Alphatron capabilities include the following:

1.) Force Measurement

   a.) Range; 0 - 220 Newtons

   b.) Resolution; .08 Newtons

2.) Displacement Measurement

   a.) Range; 1.00 Cm

   b.) Resolution; .01 Cm

3.) Displacement

   a.) Velocity Range; .005 Cm/sec to 2.5 Cm/sec

   b.) Positioning Accuracy; .00025 Cm

The Alphatron consists of three major subsystems: the

mechanical system, the control and data acquisition system, and the environmental system [see Appendix I].

Tissue Acquisition

Tissue samples of endopelvic fascia were obtained from fresh autopsy material by Dr. Richardson. The tissue was cut to a standard specimen size, approximately .5 Cm x 3.0 Cm, and weighed. Each end of the specimen was sandwiched between two small pieces of bamboo and secured with surgical thread. A hook was then inserted through each of the (bamboo and tissue) "sandwiches". The top hook was attachec to the loading rod, and the bottom hook was attached to the submergible force transducer.

The tissue was then preconditioned by loading and unloading the specimen at a fixed strain rate until the resultant hysteresis loops approached a steady state, after approximately 5-10 cycles. After preconditioning, the specimen length was adjusted so that the force transducer read zero force and the specimen showed no buckling. The tissue specimen was now said to be at resting length lo, and this length was measured to .001 CM with a cathetometer (Gaertner Scientific Corporation).

Testing Procedure

The three different tests performed on the tissue specimens were simple elongation, stress-relaxation, and ultimate strength. During each test, the temperature was maintained at a constant 37°C.

For the simple elongation tests, the specimen was first loaded at a constant strain rate to a desired final stress level. Both force and displacement were displayed and recorded by an oscillograph.

In the stress-relaxation tests, the specimen was stretched at the maximum speed of the Alphatron to a desired strain level; the decrease in force with increasing time was recorded. The long term results are for tests with a time span of 17 minutes. Short term tests were conducted to test the effect of the value of the initial stretch and the strain rate. The short term tests cover a time span of 50 seconds.

A third test was done after all other desired testing of that sample was complete. In order to find the ultimate strength of the tissue, the specimen was stretched at a constant strain rate until it failed.

CHAPTER III

Data Reduction

It is our task to unify theoretically the results of these various one-dimensional tests. A correct theoretical formulation must bring out the unity among different types of experiments; only the formulation that is consistant with results of different experiments will be useful [1].

Fung[4] has proposed a quasilinear viscoelastic model which has been applied to soft biological tissues [2,3] as well as various polyamide fibers [7]. During testing, the inertial force is approximately 3 or 4 orders of magnitude less than the stress, hence the quasi-static assumption in classical viscoelasticity is used.

The Quasilinear Viscoelastic Model

The history of the stress response, $K(t)$, for a suddenly applied strain history, $\lambda(t)=1+(\lambda-1)H(t)$, is in general a nonlinear function of many variables including stretch magnitude, λ, temperature, Θ, time, t, and the chemical composition of the bath. Thus,

$$K = K(\lambda, \Theta, \ldots t) \tag{1}$$

In this study, the temperature and chemical composition of the environmental bath was fixed. Thus, equation (1) can be rewritten as

$$K = K(\lambda, t) \tag{2}$$

and following the proposed model,

$$K(\lambda, t) = T^{(e)}(\lambda) G(t) \tag{3}$$

where $T^{(e)}(\lambda)$ is the elastic response and $G(t)$, the reduced relaxation function, is a dimensionless function of time possessing the property that $G(0)=1$.

Elastic Response. Fung[3] has introduced the use of an exponential function to describe the stress-strain behavior of biological tissues;

$$T^{(e)} = \beta e^{\alpha(\lambda-1)} - \beta \tag{4}$$

which has been used for rabbit mesentery [1,2] and the series element of myocardium [10].

History Dependent Response. Assuming that the hysteresis curves obtained from constant strain rates during loading and unloading are independent of the strain rate

[4], we choose a system with a continuous relaxation spectrum, $S(\tau)$. Then the reduced relaxation function, $G(t)$, can be written as

$$G(t) = \frac{1 + \int_0^\infty S(\tau) e^{-t/\tau} d\tau}{1 + \int_0^\infty S(\tau)\, d\tau} \tag{5}$$

$S(\tau)$ should be chosen so that $G(t)$ is compatible with experimental observations. Assuming that hysteresis curves are insensitive to strain rate, we require that the dynamic response to a sinusoidal strain input be independant of the input frequency. Following Fung[5], we choose

$$S(\tau) = C/\tau \quad \text{for } \tau_1 \le \tau \le \tau_2 \tag{6}$$

$$S(\tau) = 0 \quad \text{for } \tau < \tau_1, \tau_2 < \tau \tag{7}$$

where C is a constant. Thus,

$$G(t) = \frac{1 + C\left[E1(t/\tau_2) - E1(t/\tau_1)\right]}{1 + C \ln(\tau_2/\tau_1)} \tag{8}$$

where $E1(z)$ is the exponential integral

$$E1(z) = \int_z^\infty \frac{e^{-t}}{t}\, dt \tag{13}$$

The corresponding reduced relaxation function, $G(t)$, can be approximated in terms of $E1(z)$ for $0<z<1$.

$$E1(z) = -\gamma - \ln z - \sum_{n=1}^{\infty} \frac{(-1)^n z^n}{n\, n!} \qquad (10)$$

where γ is the Euler constant with an approximate value of

$$\gamma = 0.57721$$

and for $1 \leq z \leq \infty$, E1 can be approximated by:

$$ze\ E1(z) = \frac{z + a1z + a2}{z + a3z + a4} \qquad (11)$$

where,

$$a1 = 2.33473$$
$$a2 = 0.25062$$
$$a3 = 3.33066$$
$$a4 = 1.68153$$

and for large values of z,

$$E1(z) \sim \frac{e}{z}\left\{1 - \frac{1}{z} + \frac{2!}{z^2} - \frac{3!}{z^3} \cdots\right\} \qquad (12)$$

From this it becomes apparent that for very large z;

$$E1(z) = 0(e^{-z}/z) \qquad (13)$$

and for small values of z;

$$E1(z) = -\gamma - \ln z + 0(z) \qquad (14)$$

After some further simplification, we obtain the following two equations describing the stress relaxation, G(t), and the slope of the middle portion of the stress relaxation

function, Sr. The three parameters are then ζ_1, ζ_2, and C.

$$G(t) = \frac{1 - C\gamma + C\ln(\zeta_2) - C\ln(t)}{1 + C\ln(\zeta_2) - C\ln(\zeta_1)} \tag{15}$$

$$Sr = \frac{dG}{d(\ln t)} = \frac{-C}{1 - C\ln(\zeta_2) - C\ln(\zeta_1)} \tag{16}$$

CHAPTER IV

Results

Two tissue samples were tested in the Alphatron after preliminary tests were performed to establish the laboratory protocol. Sample 1 was from the pubourethral ligament and was composed largely of smooth muscle [9]. Sample 2 was composec of dense regular tissue from the pelvic sidewall. Both tissue specimens were obtained from the same source and were tested within six hours after harvesting.

The calculation scheme to find $\alpha, \beta, \zeta_1, \zeta_2$, and C follows the theoretical framework outlined in Chapter III.

Simple Elongation

The Eulerian stress, T, was calculated as

$$T = (F)(l_o)(\lambda) / V \qquad (17)$$

where F is the measured force; l_o is the initial reference length; λ is the stretch ratio defined as the instantaneous length, l, divided by l_o; and V is the volume of the specimen. Assuming that the specimen is prismatic and has a specific density close to the specific density of water, the volume can be determined directly from the mass of the

specimen [2]. The secant elastic modulus, $\Delta T/\Delta\lambda$, was calculated as follows;

$$\frac{\Delta Ti}{\Delta\lambda i} = \frac{Ti - Ti-1}{\lambda i - \lambda i-1} \qquad (18)$$

The stress, T, versus the stretch ratio, λ, was plotted in Figure 1. The shape of the curve suggested an exponential relationship, so T was replotted versus λ on semi-log paper, Figure 2. The result was close to a straight line. Figure 3 shows a plot of $\Delta T/\Delta\lambda$ versus T and displays a linear relationship. A linear regression was carried out on this curve to find the slope, α, the y-intercept, α/β, and r (a correlation coefficient indicating the linearity of the curve). These results are summarized in Table 1.

Stress-Relaxation

The reduced relaxation function, G(t), was calculated by first calculating the stress, T, and then normalizing the stress to find G(t);

$$G(t) = T(t)/T(0) \tag{25}$$

thus, $G(0) = 1$. The reduced relaxation function was then plotted versus time on semi-log paper, Figure 4. the constants τ_1, τ_2, and C were then calculated using the following equations;

$$C = (dG(t)/d\ln t)/G(\infty) \tag{20}$$

$$\tau_2 = e^{\left\{\frac{1}{C}\left[(G(t_0)/G(\infty)) - 1 + C\gamma + C\ln t_0\right]\right\}} \tag{21}$$

$$\tau_1 = \tau_2\left[e^{\left\{\left[G(\infty)-1\right]/\left[C\,G(\infty)\right]\right\}}\right] \tag{22}$$

where $G(\infty)$ is the asymptotic value of the relaxation function as $t \to \infty$. The constants τ_1, τ_2, and C were then substituted into equation (8) to give theoretical values of G(t) versus t. This "theoretical curve" was then plotted with the experimental values of G(t) in Figure 4. The constants τ_1, τ_2, and C are given in Table 1.

Short term relaxation tests were performed on Sample 2 to determine the effect of strain rates on the reduced relaxation function. G(t) was plotted versus time for four different strain rates in Figure 5.

Three more short term relaxation tests were performed on Sample 2 to determine the effect of initial stretch on the reduced relaxation function. G(t) was plotted versus time for three different initial stretches (Figure 6).

Ultimate Strength

No data was recorded for the ultimate strength test of Sample 1. During the test, the tissue was pulled out of the bamboo clamps before the tissue failed. While testing Sample 2, the displacement of the tissue went out of the linear range of the LVDT before the specimen failed. However, the stress, T, plotted versus λ (Figure 7) shows the value of the stress and the stretch ratio at yield for Sample 2. After the stress in the specimen peaked, the stress decreased somewhat erratically for a large range of stretch. Strands of the fiber were observed to break at different strains until the final strand was approximately three times the length of the original specimen.

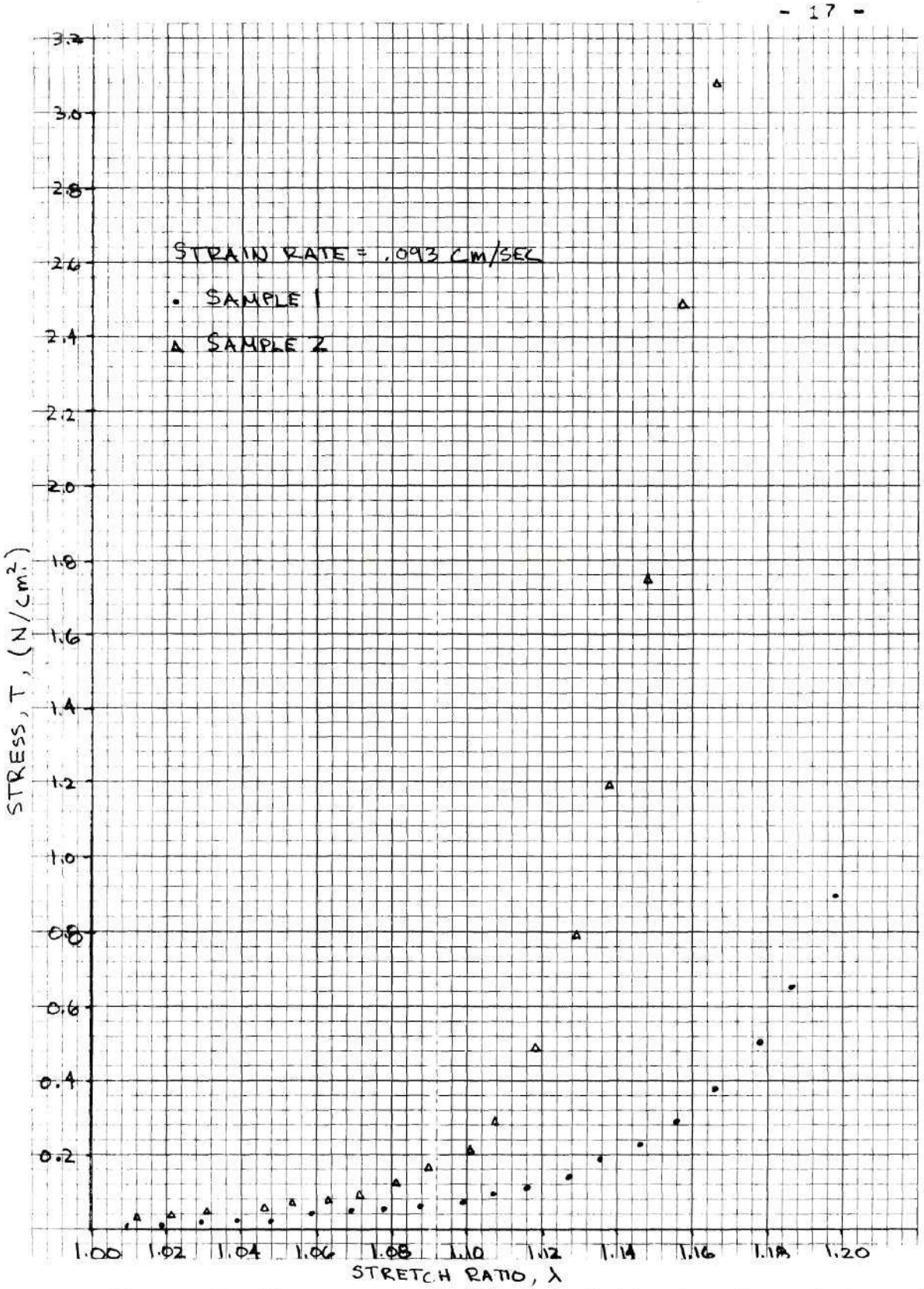

Figure 1. Stress versus Stretch Ratio (rectangular scale)

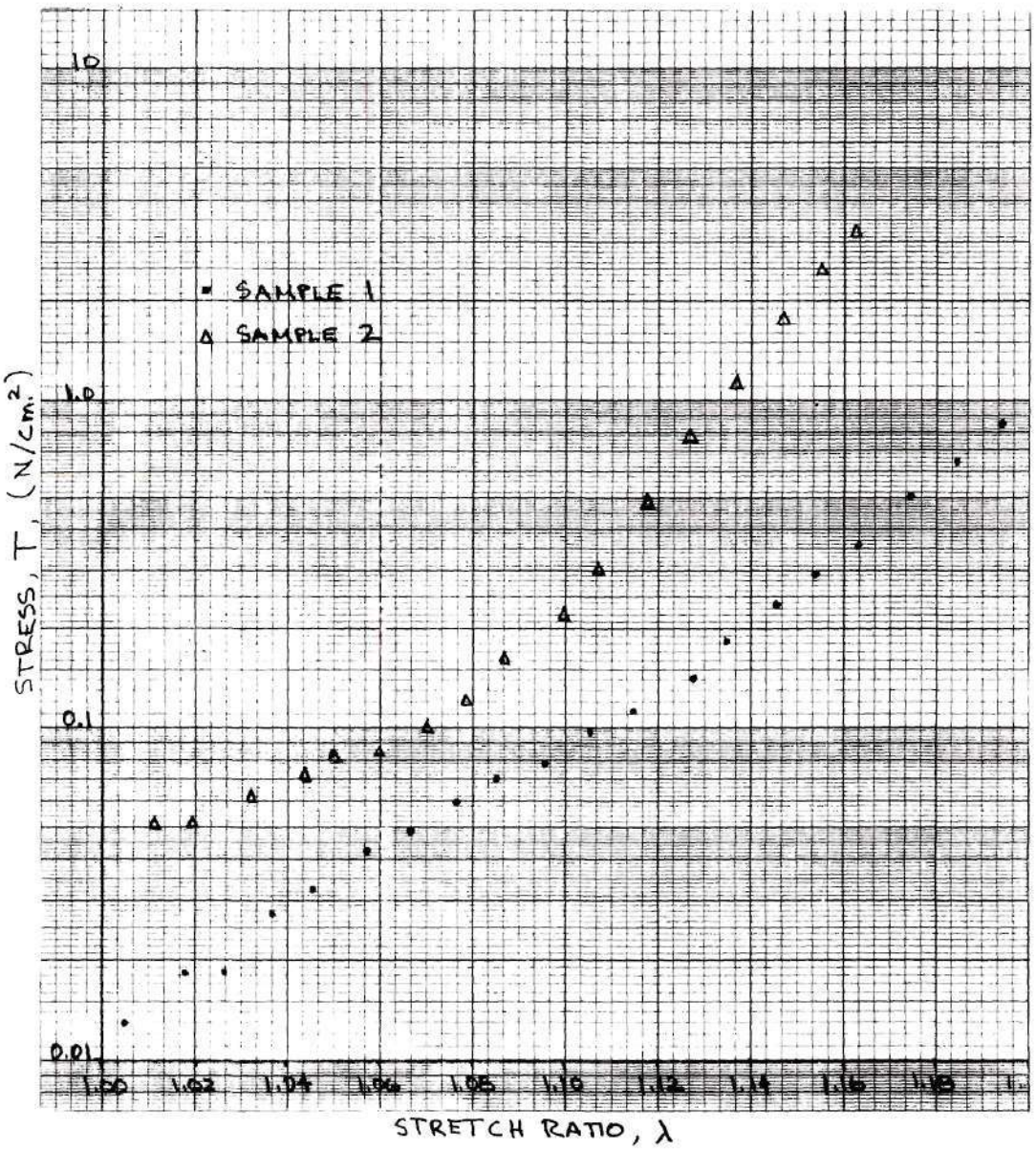

Figure 2. Stress versus Stretch Ratio (semi-log scale)

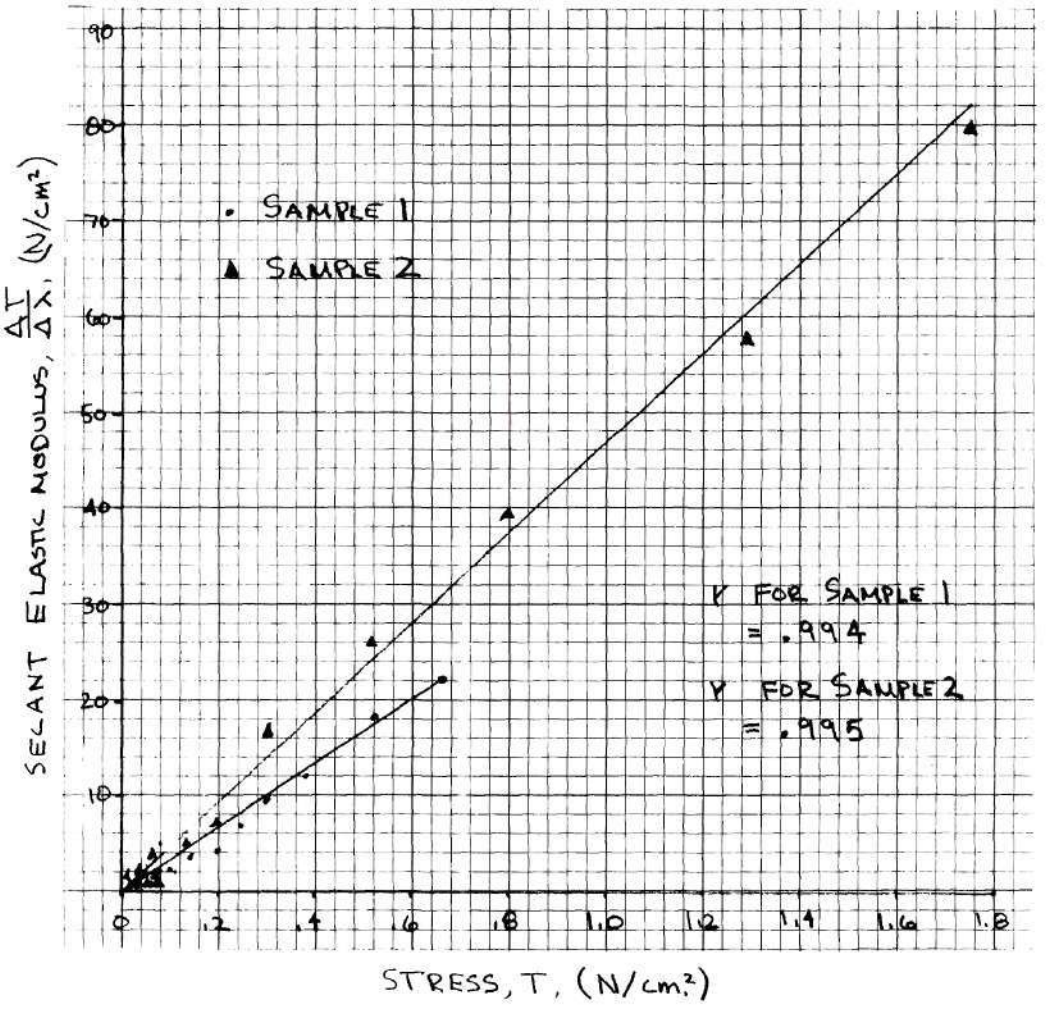

Figure 3. Secant Elastic Modulus versus Stress

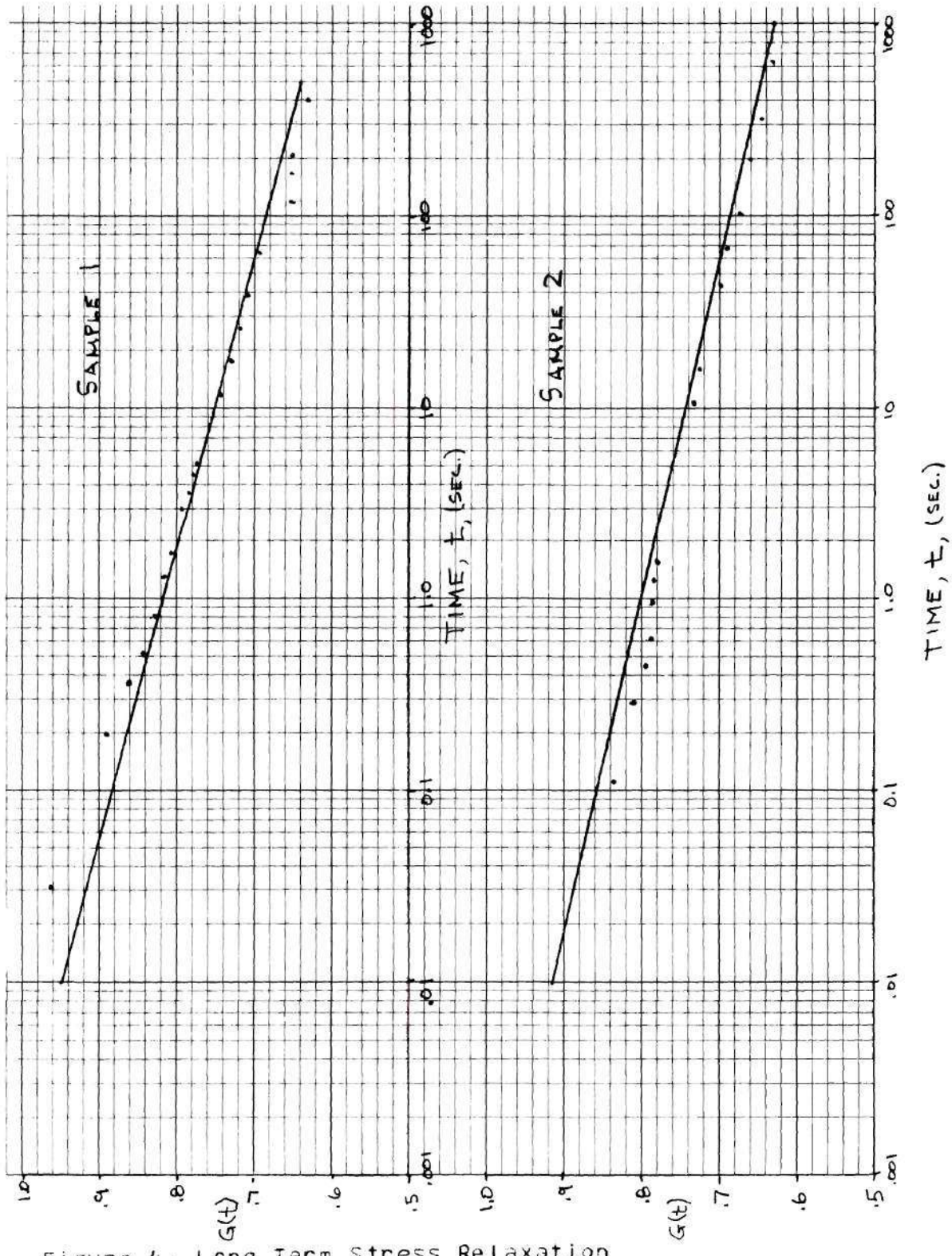

Figure 4. Long Term Stress Relaxation

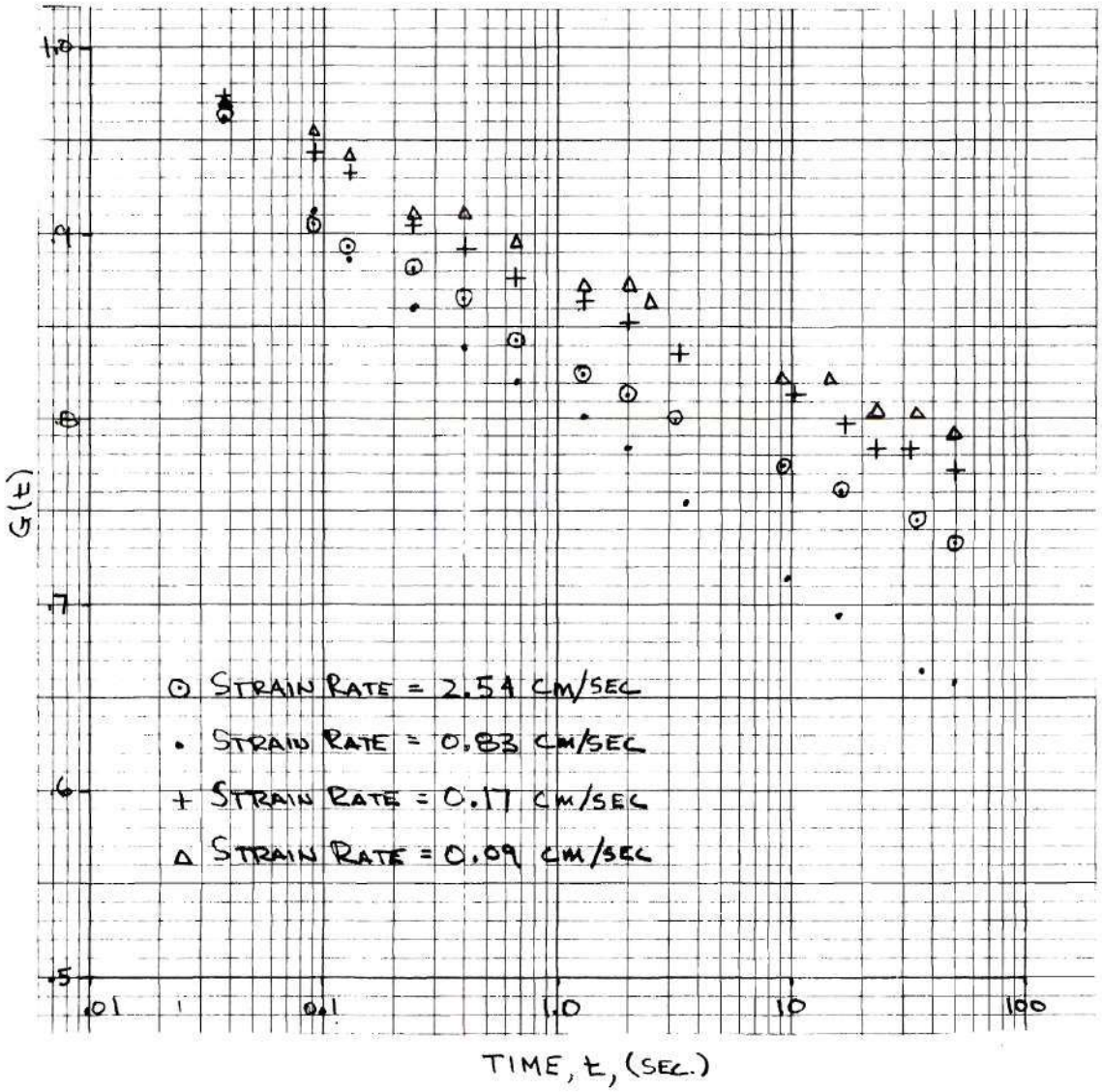

Figure 5. Effect of Strain rate on Relaxation

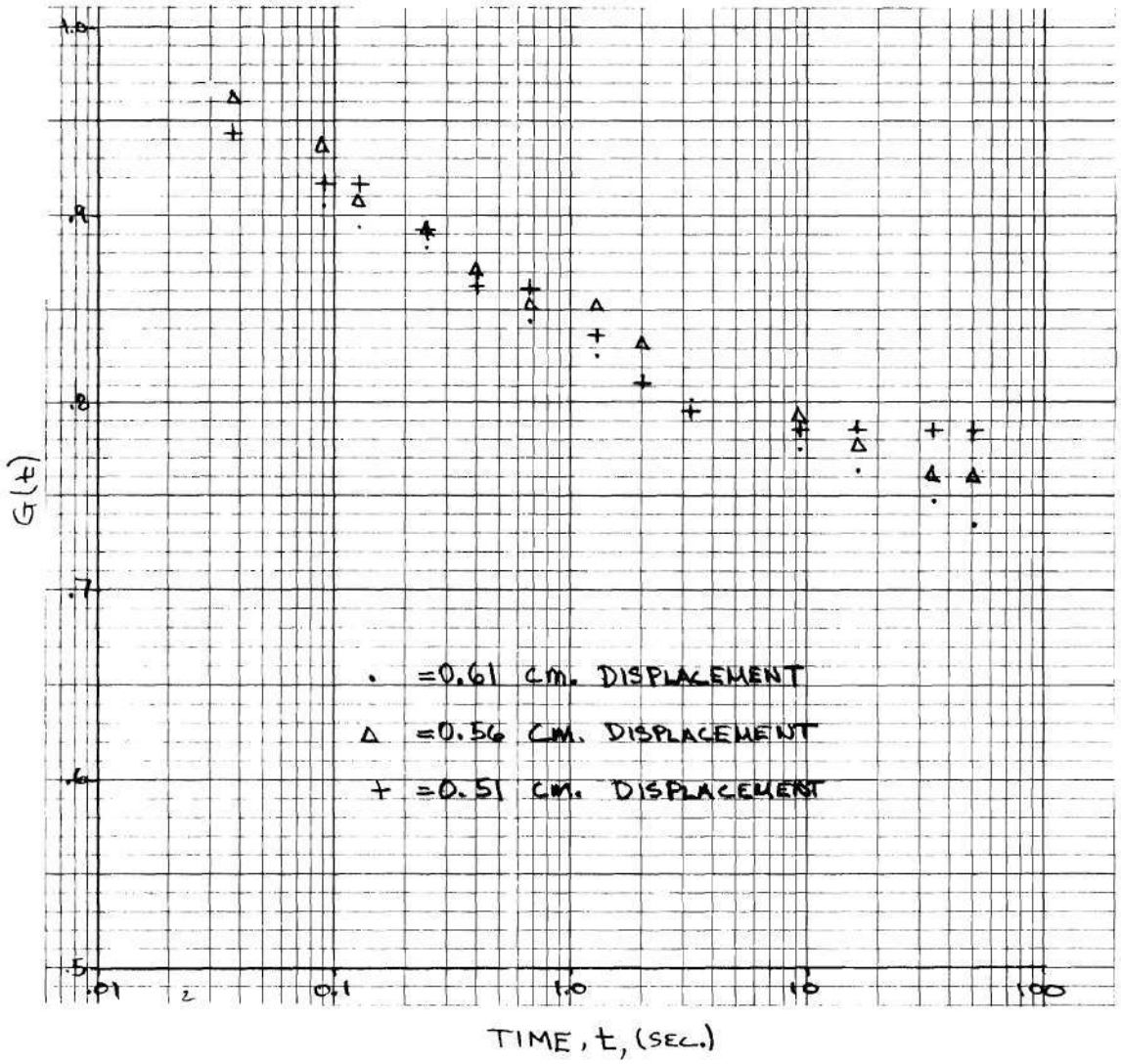

Figure 6. Effect of Initial Stretch on Relaxation

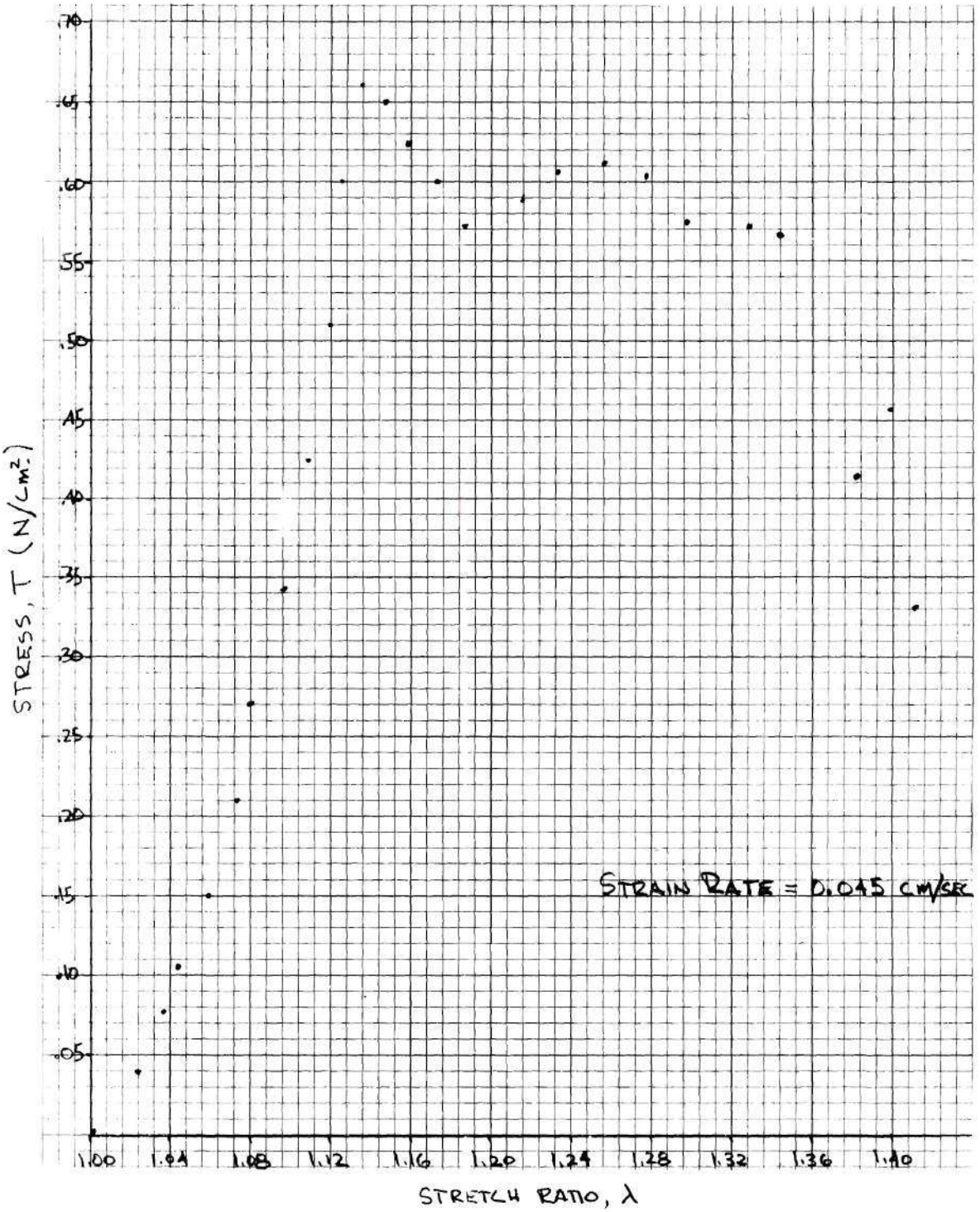

Figure 7. Ultimate Strength Test

Table 1.

Summary of Results

Simple Elongation

	α	β (N/cm)	γ
Sample 1	32.07	1.62	.994
Sample 2	47.26	4.17	.995

Stress-Relaxation

	τ_1	τ_2	C
Sample 1	.008	666.776	.0566
Sample 2	.001	1901.952	.0425

CHAPTER V

Conclusions

We have thus developed the apparatus, method of experimentation and data reduction to quantitatively observe the mechanical behavior of endopelvic fascia. The simple elongation tests show that the stress in both tissue samples increases exponentionally with stetch. Stress in the dense regular tissue from the pelvic sidewall increased faster than the stress in the pubourethral ligament- hence α_2 was greater than α_1. The long term relaxation results show that the ligament specimen relaxed faster than the connective tissue specimen. The short term relaxation tests to determine the effect of strain rates on the reduced relaxation function have a separation of 13% at 50 seconds, which seems to indicate that the strain rate is an important variable in the stress-relaxation tests. The short term relaxation tests to determine the effect of initial stretch show a separation of 5% at 50 seconds, which seems to indicate that the reduced relaxation function is relatively insensitive to the initial stretch for the range tested. These results closely resemble the results of rabbit mesentery obtained by Fung and Chen [2], hence the

quasilinear viscoelastic law is applied. The ultimate strength test seems to indicate that the dense regular connective tissue has an ability to maintain a load even after the yielding point.

However, although endopelvic fascia specimens were tested quantitatively, we have gained only a tentative insight into the behavior of the endopelvic fascia and its role in pelvic relaxation. The ultimate strength test suggests the possibility of a localized failure in the endopelvic fascia giving a large deformation of the pelvic supports and resulting in the symptoms of pelvic relaxation. However, many more specimens must be tested - tissues from different localities, histological composites, age, and case histories. Improved methods of harvesting and clampling the specimens must also be developed to insure consistant handling of the tissue and prevent failure of the clamp before the tearing of the specimen.

Further biomechanical studies coupled with the determination of geometrical configurations in the pelvis and further clinical investigations may enable an eventual structural analysis of the pelvic supports. Only then can a rational therapy be devised to treat pelvic support defects and consideration of the prevention of this problem begin.

APPENDIX I

The Alphatron

Mechanical System. The mechanical system consists of the structural frame, the lead screw assembly, and the digitally controlled D. C. Servo Motor. This system was designed by the author and assembled in the Engineering Science and Mechanics machine shop.

As shown in Figure 2, the main frame was constructed by welding two steel plates (2"x2") onto two steel I-beams (2"x8"x4") with two aluminum platforms bolted to the I-beams to provide support of the lead screw assembly and the environmental chamber. A lead screw with a double preloaded nut (Beaver Precision) was mounted between two bearings (Sealmaster) and bolted to the frame in such a way that the lead screw nut had 1.25" of vertical travel in each direction. A steel plate (lobe plate) was constructed and mounted onto the lead screw nut. Two stainless steel guide rods were mounted to the frame, and inserted through two Teflon bearings on the lobe plate. Thus, rotation of the lead screw resulted in vertical motion of the lobe plate. From the front end of the lobe plate, a stainless steel loading rod was extended to the tissue specimen. A LVDT

(linear variable displacement transformer) rod was mounted to the back end of the lobe nut. The top of the lead screw was coupled (Pic Design) directly to the motor (Torque Systems).

Control And Data Acquisition System. The control and data acquisition system is a special purpose digital processing unit designed to provide control of the mechanical system and to monitor and record data from the force and displacement transducers. The system was designed by John L. Lee and assembled by Steve Lyons and Steve Benoist.

Control of the experimental system is provided by a M6800 (Motorola) micro-computer system. A PIA (periferal interface acapter) is used to provide a parallel interface to the digitally controlled analog servo system (Torque Systems) via a controlling junction box. A pulse train is generated by the micro-computer and sent to the digital logic ccntained in the Torque System device. The Torque System device is set to operate in the PLO (Phase Lock Operation) mode in which the motor shaft is incremented by .36 For each input pulse. Another input line from the micro-computer is set at "0" for counter clockwise rotation, or set at "1" for clockwise rotation. Thus, input to the

micro-computer for direction, frequency and duration of the pulse train dictates the magnitude, direction, and velocity of the motor rotation. The Torque System device achieves the desired rotation utilizing its own internal compensation circuitry, optical encoder, tachometer, and digital to analog converters.

Force data is provided by either a submergible force transducer (Kistler-Morse) mounted below the tissue specimen, or by a Statham Gold cell with an adapter. The adapters have upper load limits ranging from 2 pounds to 50 pounds. Displacement data is obtained from an LVDT (Trans-Tek). Voltage from the force transducer is amplified with a Validyne strain gage amplifier. A Validyne carrier demodulator plug-in unit sends a carrier wave to the LVDT where the signal is modulated by the LVDT and the resultant signal is then demodulated and amplified by the Validyne. Thus, the Validyne output consists of two voltages- one corresponding to force, the other corresponding to position. The two voltages are then displayed and recorded by an oscillograph (Honeywell Visicorder).

Environmental System. The purpose of the environmental system is to maintain the tissue in a constant

and consistant chemical mileau during the in vitro testing. The environmental tank, featuring a mobile force transducer platform, was designed by the author and assembled in the Nuclear Engineering department's machine shop by W. B. Jeter.

The tissue was immersed in a bath of Ringer's solution which provided control of the temperature and maintained a constant chemical environment. Temperature of the bath was maintained by a thermostat and pump (Bromville Scientific Incorporated) which circulated water at a constant temperature around the specimen tank.

Experimental Method

The Alphatron can be operated in one of two modes: manual or computer controlled. In the manual mode the controlling junction box was set for manual operation. A frequency generator was interfaced with the controlling junction box and regulated the speed of motor rotation. The result was a continuously variable strain rate. One switch on the junction box determined the direction of motor rotation (up or down direction of the lobe plate movement), and another switch on the junction box was for starting or stopping the motor rotation. In addition to the switches on

the junction box, two safety switches were mounted to the Alphatron to stop the motor and prevent the lobe plate from crashing into the structural plates above or below.

In the computer controlled mode, the junction box switches were set for computer operation. The M6800 was then programmed to control the time course of motor rotation by generating a pulse train which replaced the need for a function generator. In the computer controlled mode the safety switches were also operational.

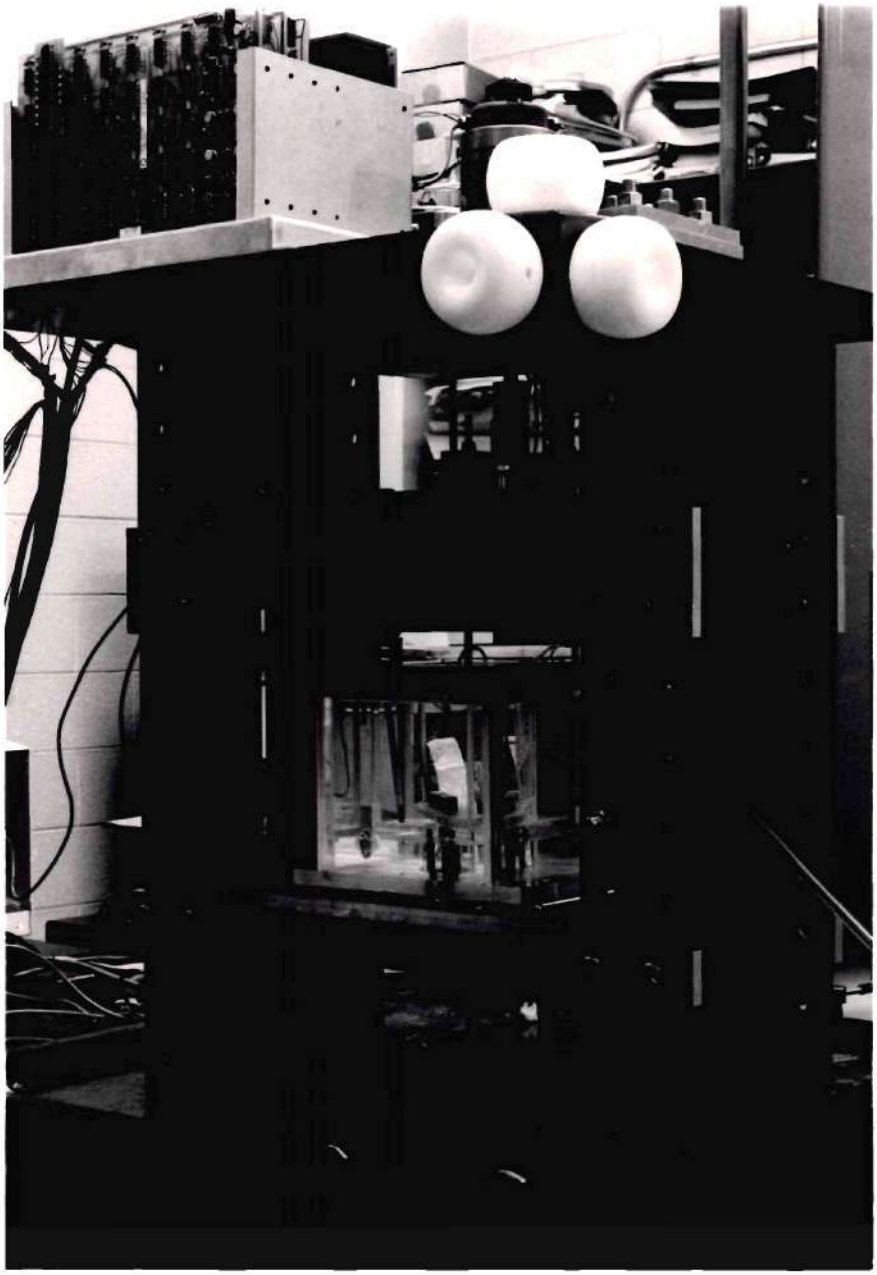

Figure 8. The Alphatron (oblique)

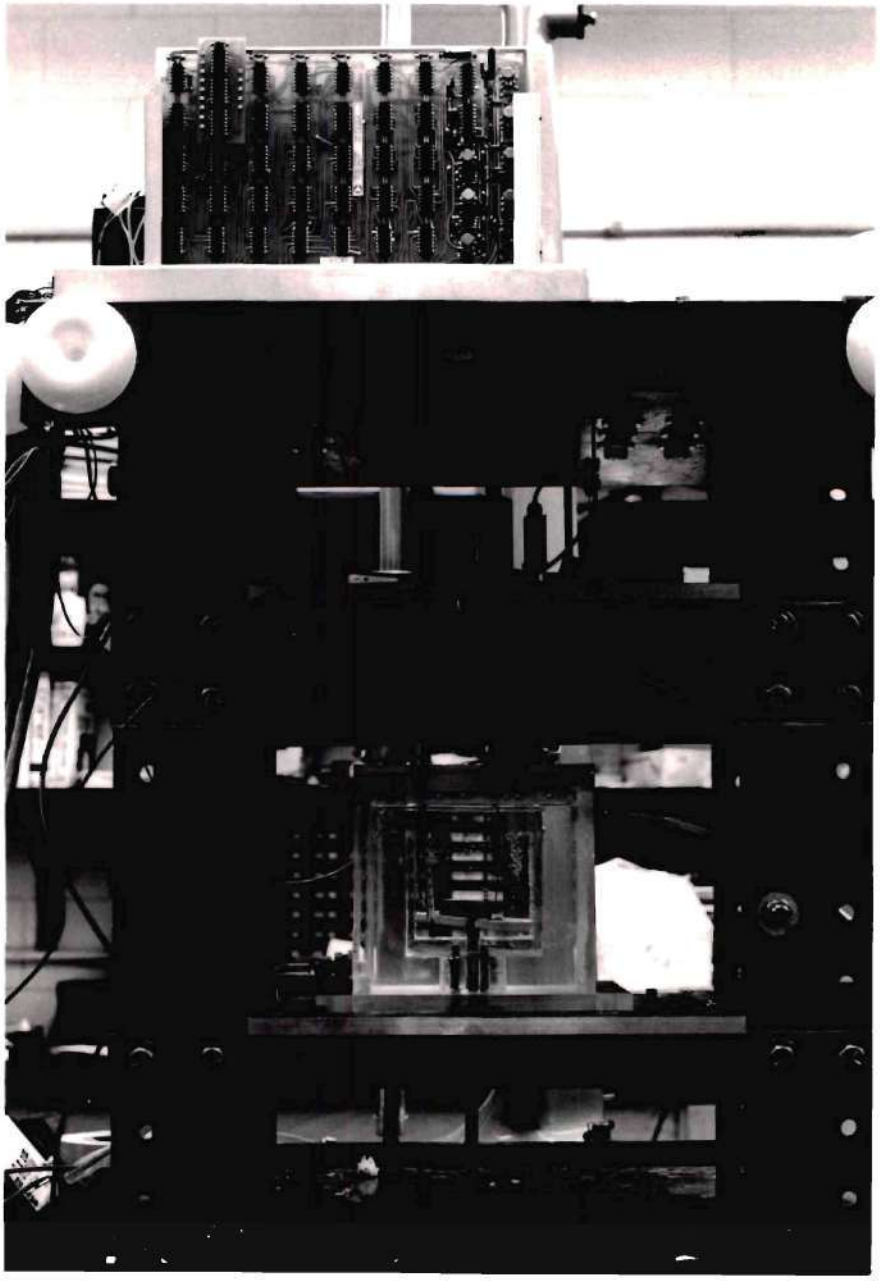

Figure 9. The Alphatron (front)

Figure 10. The Alphatron (back)

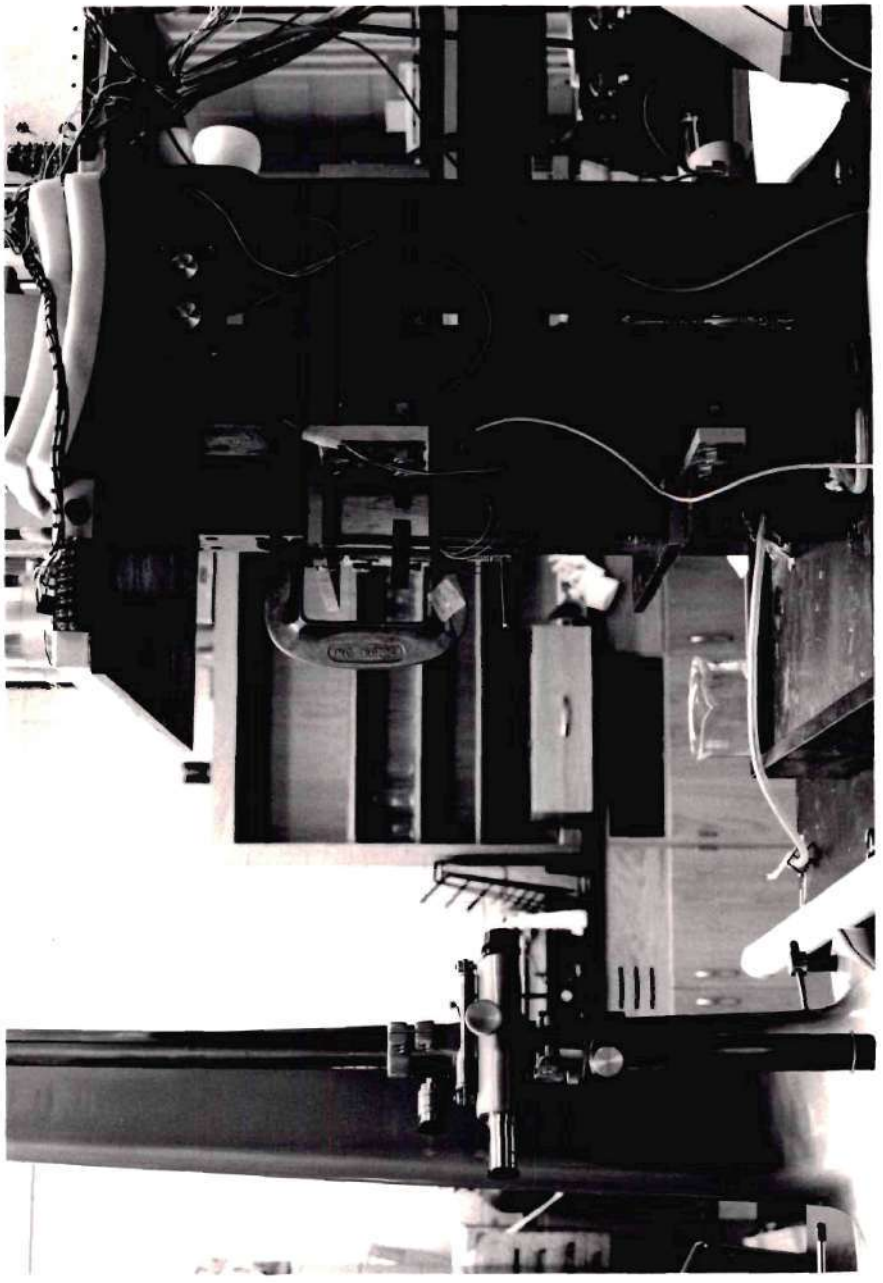

Figure 11. The Alphatron (side)

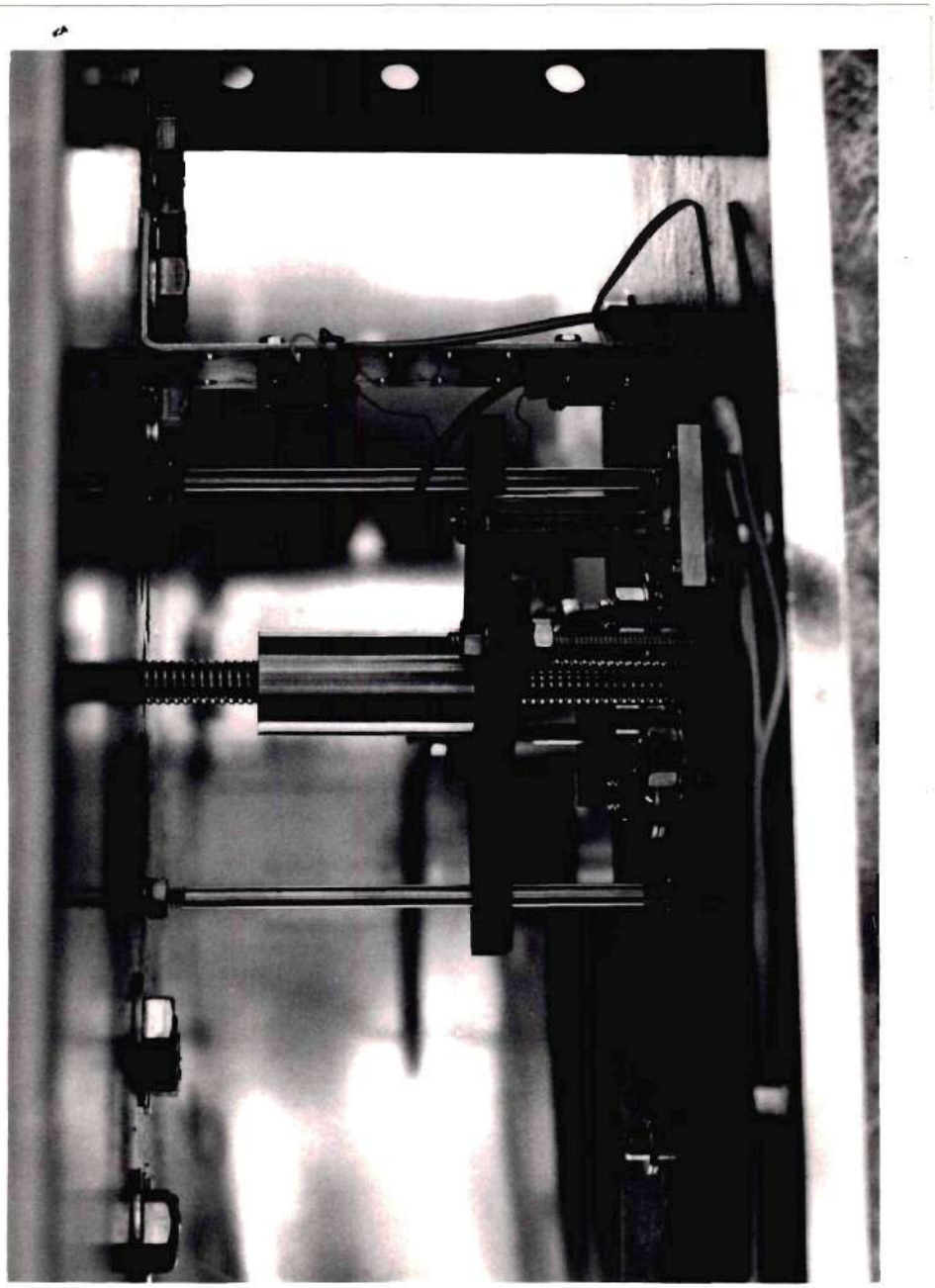

Figure 12. Lobe Plate and Lead Screw

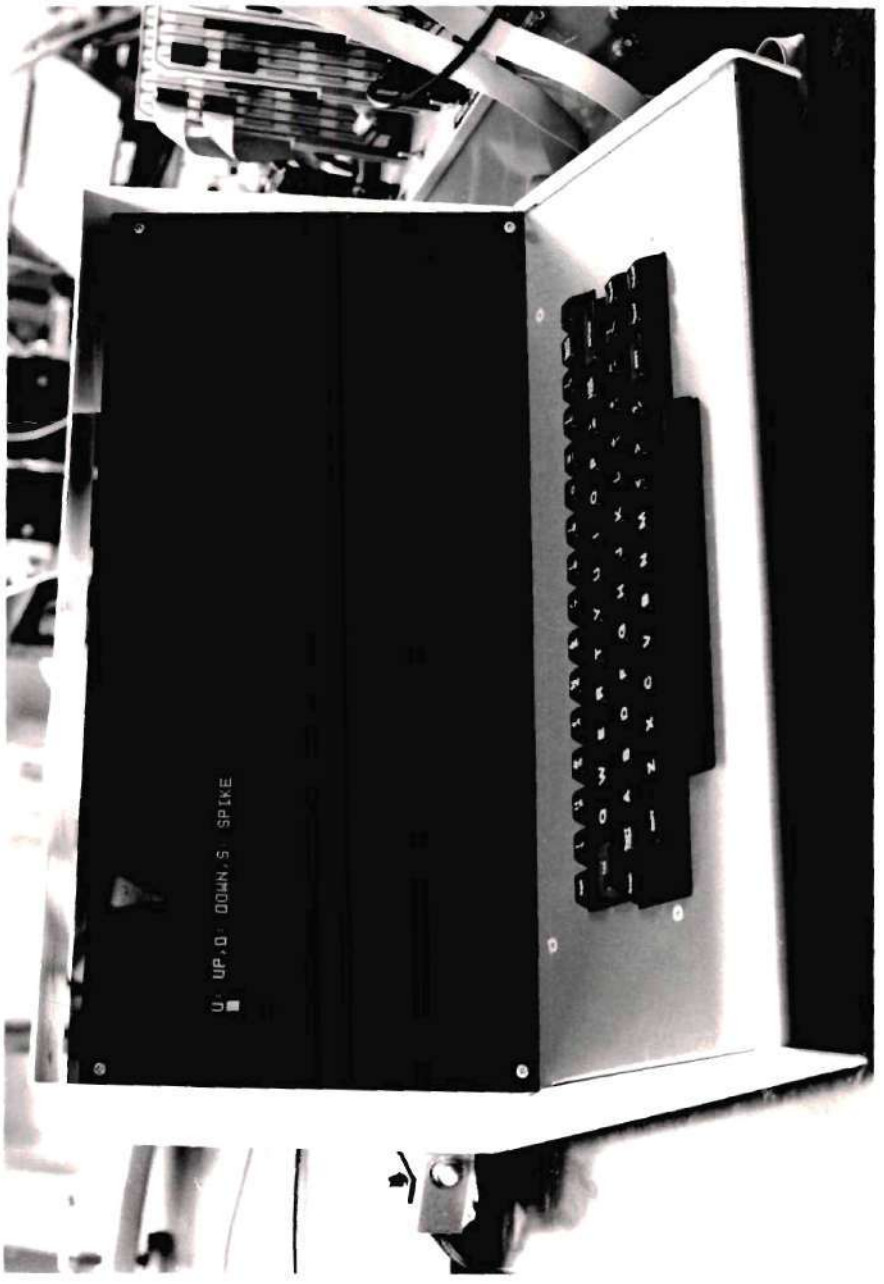

Figure 13. The Micro-Computer

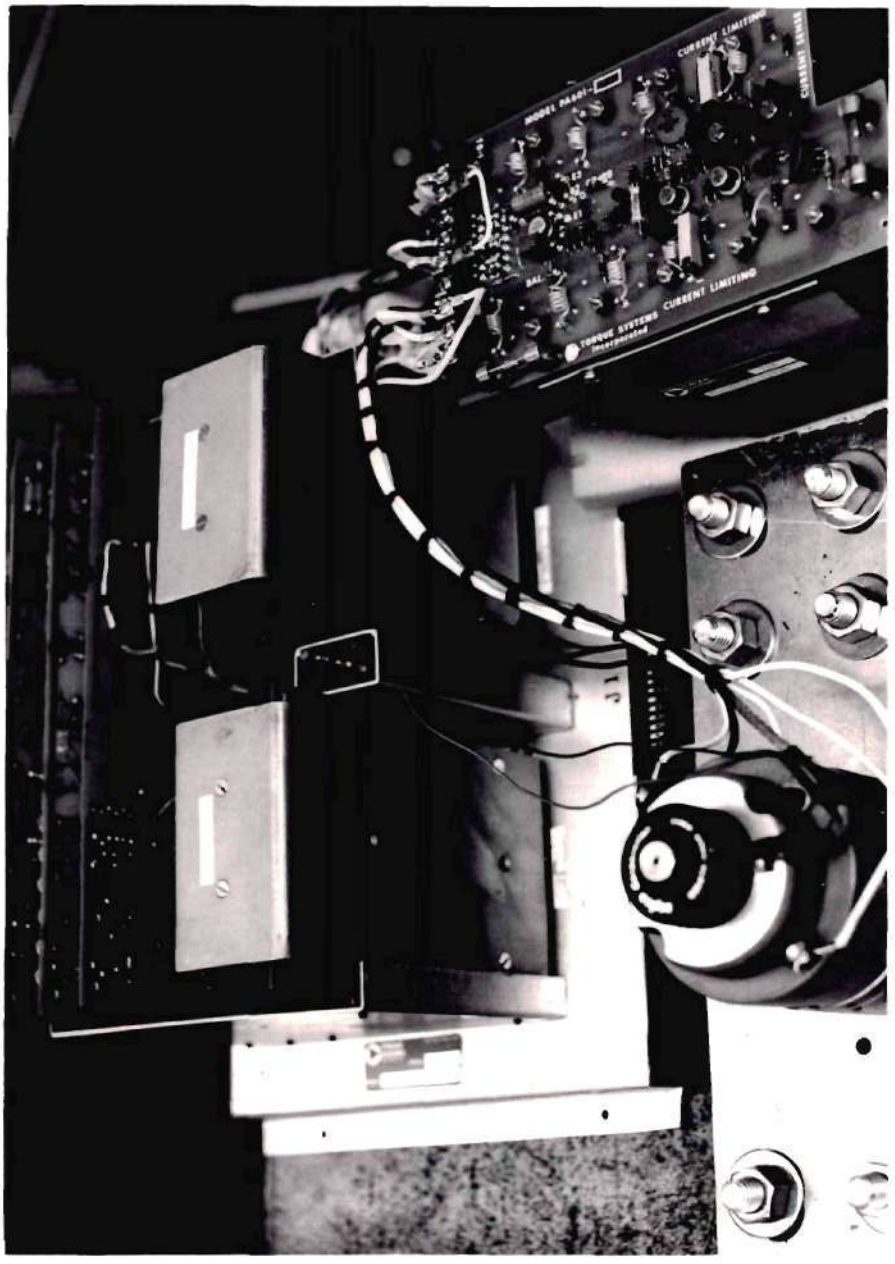

Figure 14. The Torque System Device

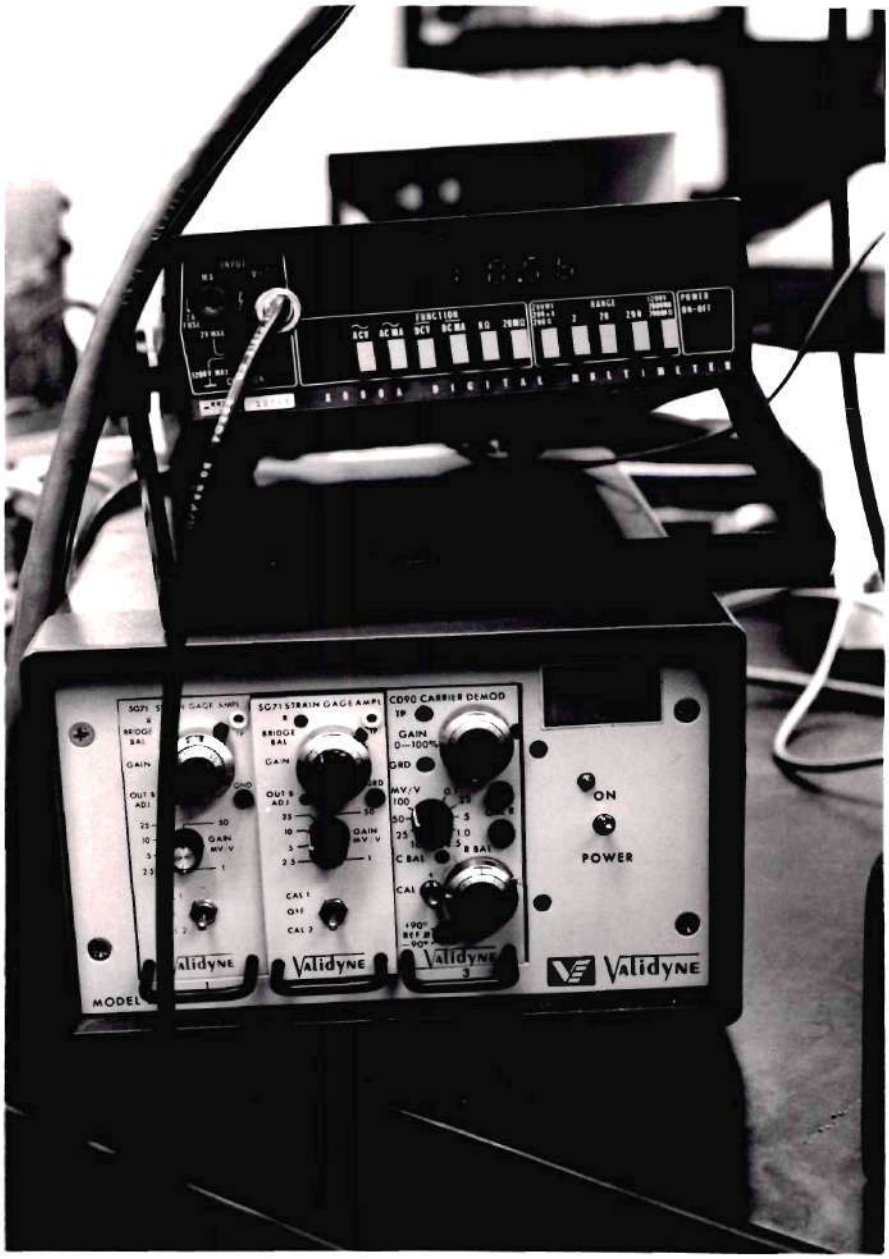

Figure 15. The Validyne Amplifier

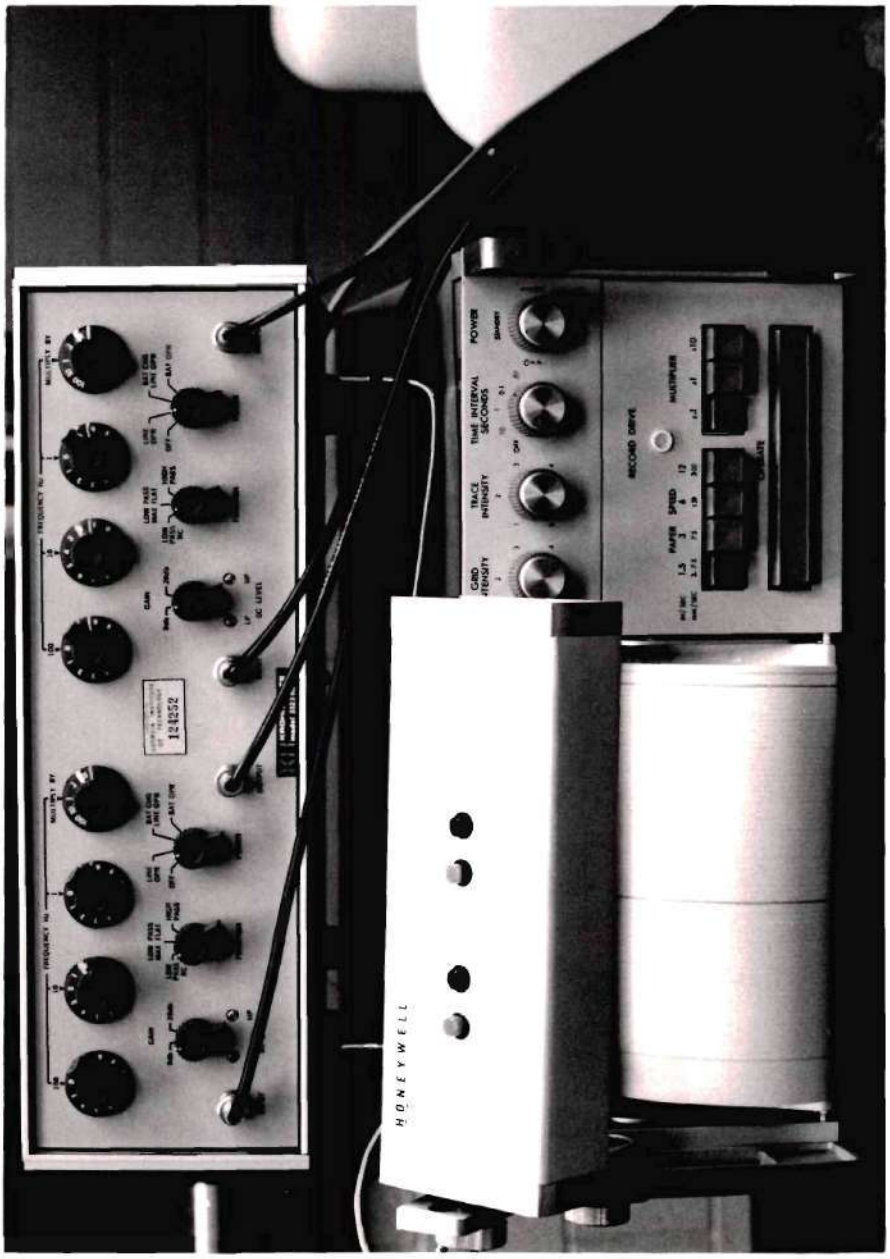

Figure 16. The Oscillograph

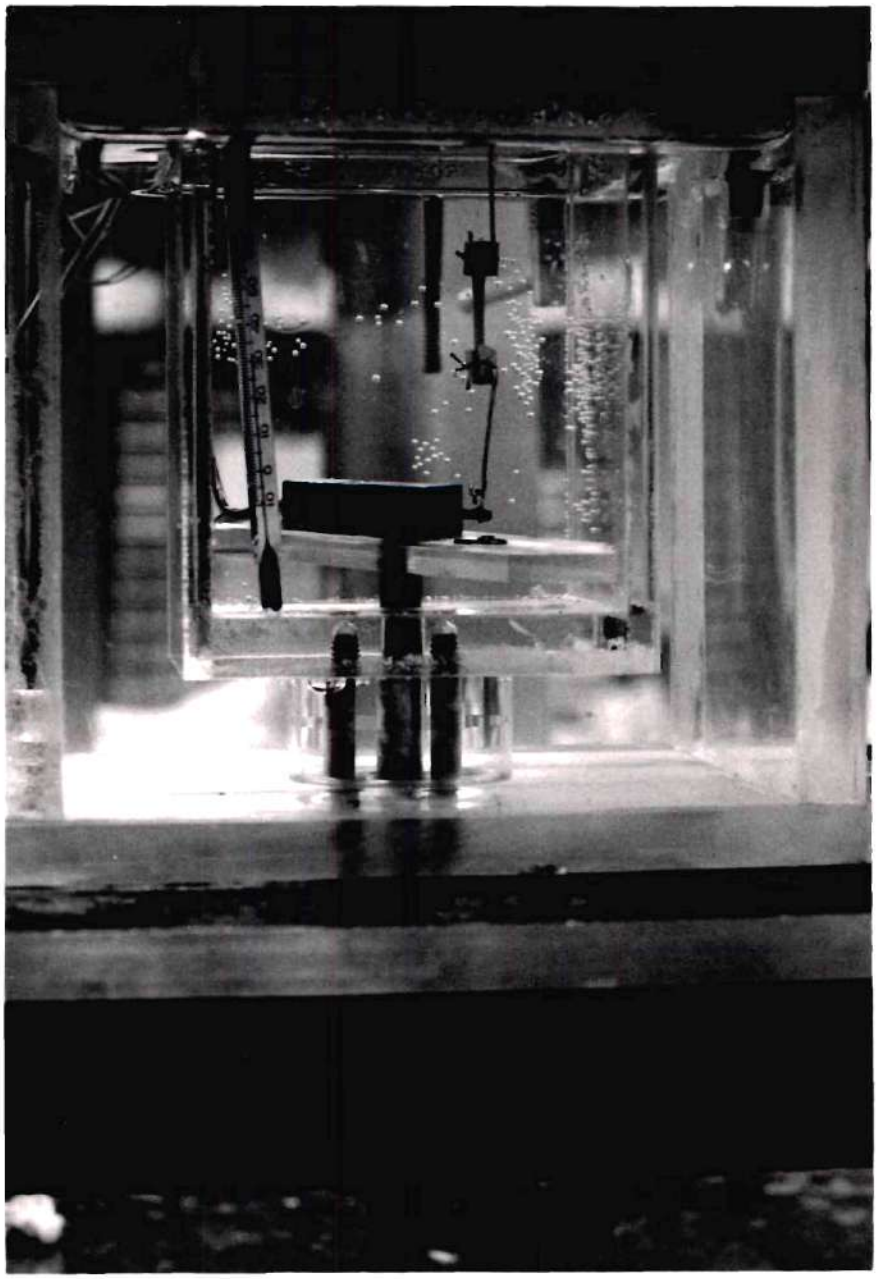

Figure 17. The Environmental Chamber and Specimen

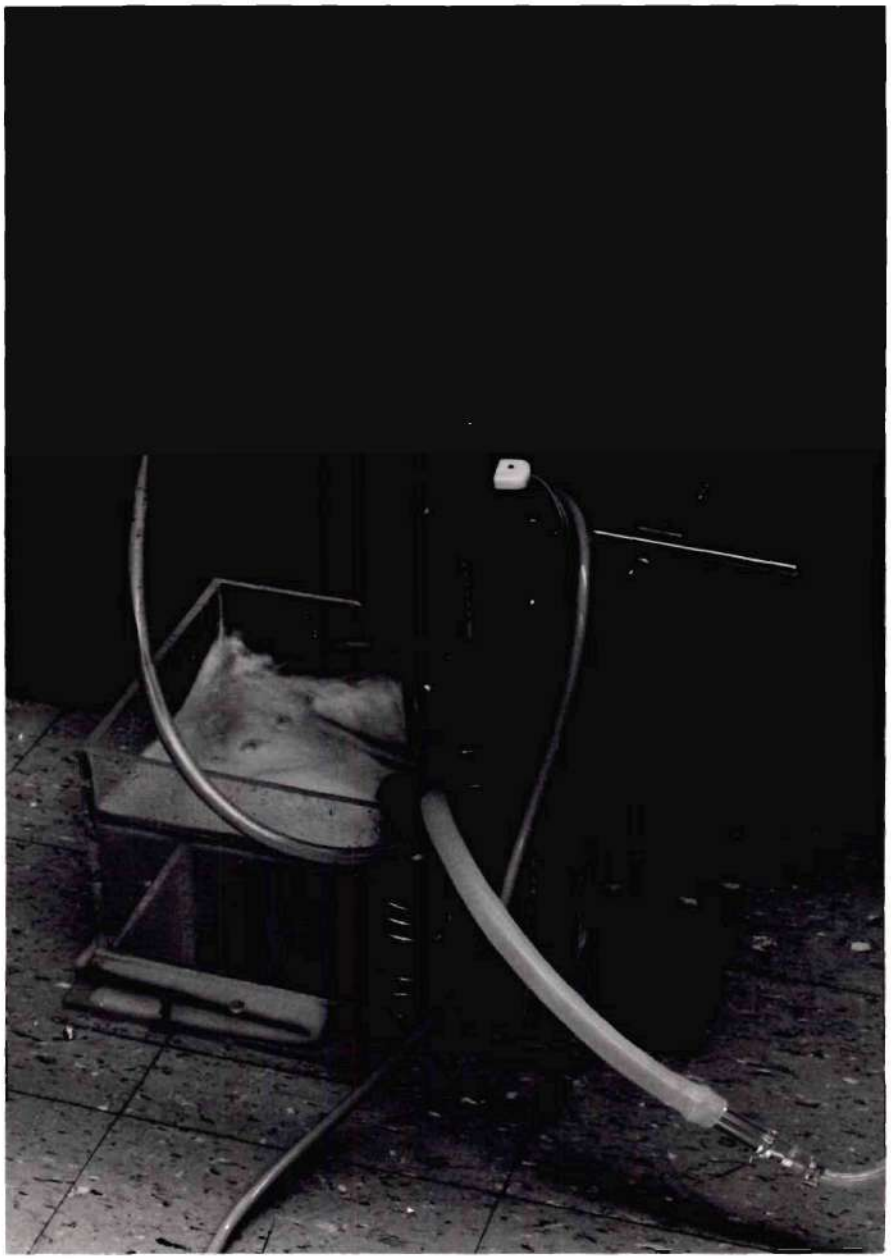

Figure 18. The Thermostat and Pump Unit

APPENDIX II

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